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DEVELOPMENT OF A BLOOD-PRESSURE TRANSDUCER

Prepared for:

NATIONAL AERONAUTICS AND SPACE ADMINISTRATION
AMES RESEARCH CENTER
MOFFETT FIELD, CALIFORNIA

CONTRACT NAS 2-809

By: G. L. Pressman P. M. Newgard

STANFORD RESEARCH INSTITUTE

MENLO PARK, CALIFORNIA

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ABSTRACT

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An external, arterial-blood-pressure transducer was designed, built, and tested on Contract NAS 2-809. This transducer was an improvement over an earlier version, designed and built at SRI under Contract NAS 2-515.

The improved model provided a reasonable compromise between the small size that was desired and ease of construction. Tests showed satisfactory comparison between the reading of blood pressure from the transducer and the value given by a sphygmomanometer, thus verifying the design theory. Because of the relatively large size of the present transducer, it is most valuable for use on the radial artery.

AUTHOR

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DEVELOPMENT OF A BLOOD-PRESSURE TRANSDUCER

I INTRODUCTION

The previous research project* resulted in the derivation of a method of approach to the design of an external arterial blood-pressure transducer. As a test of this method of approach, a transducer was built and found to substantiate the basic theory, although the device suffered from several difficulties that severely restricted its usefulness. The most apparent of these were excessive zero shift and large arterial-rider size.

The main object of work under the present contract was to design and build an improved transducer that avoided some of the difficulties encountered with the earlier model and could be used for further tests of the basic theory.

The original transducer (designated Model 4) was further tested to establish the cause of its excessive zero shift. It was determined that this was primarily a temperature sensitivity effect. The transducer temperature sensitivity was found to be about 2 mv per degree centigrade, compared to a pressure sensitivity of 1 mv per 100 mmHg. Simultaneous measurement of skin temperature and transducer zero shift showed close correlation. From these tests it was concluded that temperature sensitivity must be reduced by at least an order of magnitude.

In addition, two attempts were made to record intra-arterial pressures for direct comparison with the Model 4 transducer constructed under Contract NAS 2-515. Arrangements were made at the Stanford Hospital to measure blood pressure at the radial artery of a patient during cardiac catheterization. The recording of pressure at the brachial artery is a normal procedure during this operation, and permission was given to transfer a portion of brachial artery recordings

*Contract NAS 2-515, with NASA's Ames Research Center, Moffett Field, California.

to the SRI recorder for comparison purposes. It was decided to apply the transducer to the arm that was catheterized in order to obtain the greatest proximity of the two measurements.

A number of difficulties prevented a satisfactory comparison. First, both patients (women) had faint pulse, an indication of small artery size, and it was not possible to obtain a calibrated signal. Second, catheterization of the brachial artery resulted in arterial spasm, occluding the artery and eradicating the radial artery pulse. Although the pulse eventually returned, the initial level was never regained. Finally, it was not possible to readjust the transducer position because the transducer was located in a sterile area and required complete covering with protective cloths. Although some of the difficulties could be avoided by using the uncatheterized arm, the problem of scheduling tests at the hospital led to the decision that additional tests should be delayed.

Experience with the Model 4 transducer indicated that the arterial rider was too large to permit successful application on any but the very largest of superficial arteries. It was concluded that the arterial rider should ideally be about 0.020 inch in diameter or smaller for eventual application to an artery of the temporal group.

The requirement for small rider area with the resultant low level of force, coupled with a necessity for very low displacements, creates a force-measurement problem of extreme difficulty. Adequate sensitivity was attained in the first transducer by using semiconductor strain gages. This was accomplished, however, at the expense of large rider size. At the time that transducer was constructed, the best strain gages available had a gage length of 0.5 inch. Gage length ultimately dictates all critical dimensions of the transducer for a given signal-to-noise output ratio.

During the past year, strain-gage technology advanced to the point where very small gages (0.050 inch gage length) became commercially available. This reduction in gage length allows a significant reduction

in the arterial rider size; however, as the rider area is decreased, the skin surface which contacts the rider is presented through an aperture smaller than the artery diameter. When this occurs, the deflection of the skin through the aperture (with the rider absent) becomes significantly less than the free deflection of the skin over the artery. In order to satisfy the requirements for a direct-force transducer, the arterial rider must restrain the artery to one-tenth of the deflection observed through the aperture; therefore, as the rider size is decreased, the rigidity of the transducer measuring system would have to be increased. This increase in measuring system stiffness ultimately limits the extent of reduction in rider size because the deflection becomes so small that the rider movement is not measurable.

The general problem of measuring a small force with near-zero displacement has been solved in other applications through the use of a feedback force balance system. This has an inherent advantage in that the basic position sensor need not be a linear element. It need only be extremely sensitive and possess an inherently stable zero or reference position. One form of this device, using a Hall-effect transducer was constructed as a laboratory bench model. This device displayed the basic sensitivity and stability required, but the construction of a working blood-pressure transducer of the size desired would have required much development work on the basic transducer that would not fit within the scope of the present contract.

After a review of the difficulties attendant to construction of the ideal transducer we have envisioned, it was decided that the basic purpose of this contract could best be fulfilled by completing a transducer that would provide a reasonable compromise between desired small size and construction capabilities. Using the smallest semiconductor strain gages available would allow the design of a transducer that should fit an artery such as the radial artery of a majority of people. At the same time, the construction techniques are fairly well established, and the rider aperture would not affect the free artery deflection. This transducer would then be used to check the basic

theory by testing its ability to obtain a precalibrated (calibrated prior to application to the artery) arterial blood-pressure measurement from the radial artery.

II SUMMARY

Tests of the improved arterial blood-pressure transducer show an increase in sensitivity and linearity over the model developed on the previous contract.

Temperature effect on the zero pressure output of the transducer is about 1 percent of full scale per degree centigrade; calibration sensitivity to temperature is 1.7 percent per degree centigrade. This is a definite improvement over Model 4, and temperature sensitivity is not a serious problem with the improved transducer. Until more transducers of this design are built, it will not be certain whether this greatly improved temperature response is truly representative of the design or is due to a fortuitous combination of strain gages.

The improved transducer shows no evidence of hysteresis due to friction as long as the clearance between rider and side plates remains free of obstruction. It was found necessary to apply a 0.002-inch Mylar tape cover to ensure that foreign material did not enter this clearance area.

The improved transducer exhibited a creep hysteresis with about a 60-second response time. This is believed to result from creep of the epoxy adhesive rather than from metal components. The response could be caused either by a gradual creep or relaxation of the gages on the beam, or of the beam on its mounts. Tests on bench model installations would be required to isolate the problem.

Comparison tests using the sphygmomanometer as a standard have shown that the indirect transducer gives a precalibrated measure of blood pressure when applied to the radial artery of most subjects. This indicates that the 0.030-inch rider width is a reasonable size match with the lumen of most radial arteries. In addition, it lends support to the thesis that the general method of approach to external blood-pressure measurement used in the design of this transducer is valid.

An attempt was made to obtain measurements of blood pressure from several superficial arteries. Only the radial and femoral arteries provided readings which agreed with the calibration. The small signals from the other arteries were due to one of two factors: inadequate flattening of the artery wall (caused by excessive artery depth or rigidity of overlying tissue), or insufficient coverage of the rider (caused by small artery size).

The centrifuge test emphasized the extreme difficulty in maintaining position of the transducer over an artery. This remains one of the most difficult problems to be solved before this method can be applied under conditions of acceleration.

During all tests of this transducer, the reading of blood pressure was never significantly higher than the value given by a sphgmomanometer. This result confirms the theory of direct-force measurement,¹ and justifies the use of a condition of maximum pulse height to establish correct transducer position.

The primary conclusion resulting from this work has been that indirect blood-pressure measurement by the direct-force method is valid, and that a reduction in rider area will allow the use of this method on smaller arteries such as those of the temporal group.

¹Pressman, G. L., and Newgard, P. M., "A Transducer for the Continuous External Measurement of Arterial Blood Pressure," Final Report, Contract NAS 2-515, Stanford Research Institute, Menlo Park, California (December 1961).

III DESIGN OF TRANSDUCER

Preliminary calculations showed that a transducer rider width of 0.030 inch and length of 0.125 inch would be a reasonable size to fit the radial artery, and would allow the use of available semiconductor strain gages. A free skin deflection of 100 microinches was assumed to exist due to a pulse pressure amplitude of 40 mmHg.

Micro Systems semiconductor strain gages No. MS-301-120 were chosen as representative of the state of the art in small gages. These gages have a nominal gage factor of 110, a resistance of 120 ohms, and an active gage length of 0.050 inch. Their temperature sensitivity is about 1 microinch/inch/°F. The temperature sensitivity of the gages was compensated for by using a four-arm strain-gage bridge and by including a thermistor (VECO-31 A-7) bonded to the strain beam to allow external compensation as required.

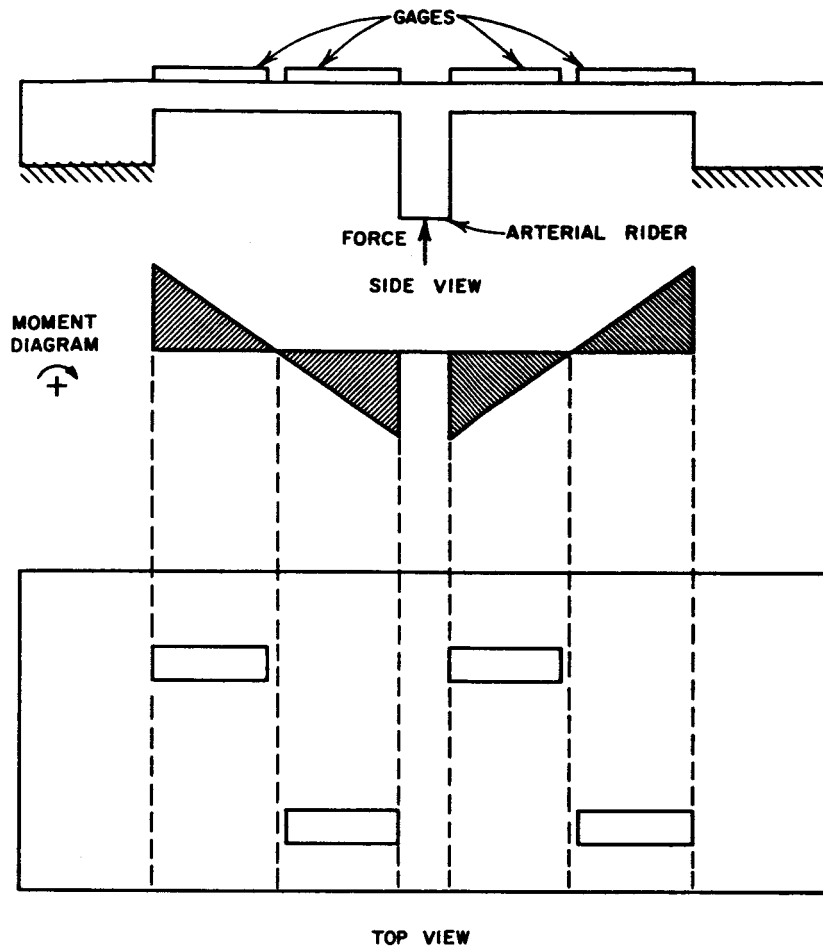
A double cantilever beam configuration was chosen (Fig. 1) to allow placing all four gages on one surface, and to facilitate construction. Figure 1 also shows the moment diagram resulting from arterial pressure force acting on the arterial rider. The four strain gages were positioned to receive the maximum average strain available in both tension and compression. Beam length and width were chosen to accommodate the available gages.

The stiffness of a double cantilever beam can be expressed in terms of beam width, W; length, L; thickness, t; and the Young's modulus of the material, E, as follows:

$$K_1 = 16 EW \left[\frac{t}{L} \right]^3 .$$

An equivalent arterial stiffness can be defined in terms of arterial pulse pressure, P; the free skin deflection resulting from the pulse pressure, d; and the arterial rider area, A:

$$K_2 = \frac{PA}{d} .$$



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FIG. 1 STRAIN-GAGE LAYOUT

Using the design criterion developed previously,¹ that the transducer stiffness should be approximately ten times greater than the physiological system, the two relations can be combined to give:

$$\frac{t}{L} = \left[\frac{5}{8} \frac{PA}{EWD} \right]^{1/3} .$$

¹Final Report on Contract NAS 2-515.

The beam dimensions we have chosen give an area $A = 3.75 \times 10^{-3}$ square inches. Free-skin deflection of the radial artery has been measured at about 100×10^{-6} inches. With

Pulse pressure	=	2 psi
Young's modulus (aluminum)	=	10×10^6 psi
Beam length	=	0.220 inch
Beam width	=	0.125 inch

the beam thickness required is

$$t = 0.220 \left[\frac{5 \times 2 \times 3.75 \times 10^{-3}}{8 \times 10 \times 10^6 \times 0.125 \times 100 \times 10^{-6}} \right]^{1/3} = 7.37 \times 10^{-3} \text{ inches.}$$

With the strain-gage configuration shown in Fig. 1, the average strain imposed on each gage will be about 1/2 of maximum strain. Using this approximation, the strain level experienced by each gage is given by the relation:

$$\text{Effective strain, } \epsilon = 0.514 \left[\frac{PA}{EW} \right]^{1/3} \frac{(d)^{2/3}}{L}$$

Using the same values for P, A, E, W, L, d as in the previous calculation results in:

$$\epsilon = 9.2 \text{ microinch per inch for 2-psi arterial pulse pressure,} \\ \text{or } 0.089 \text{ microinch per inch per mmHg.}$$

When measured by a strain-gage bridge with four active elements and 2.5 volts of bridge excitation, the resulting voltage sensitivity (E_o) should be

$$E_o = \frac{V}{4} (N) (GF) \epsilon$$

$$E_o = \frac{2.5}{4} (4) (110) (0.089) \times 10^{-6} = 24.5 \times 10^{-6} \text{ volts per mmHg}$$

where V = excitation voltage, N = number of active gages, and GF = gage factor.

Further description of design and construction is included in the Appendix.

IV TRANSDUCER CALIBRATION AND TESTING

A. STATIC CALIBRATION

A rubber membrane (0.010 inch thick) covering a pressurized cavity supplied a test pressure to the transducer for calibration. Although this apparatus is by no means a true model of the human artery, it satisfies the basic requirement that the transducer should respond only to pressure changes to the exclusion of membrane characteristics.

The static calibration curve (Fig. 2) was obtained using the apparatus described above. From Fig. 2 it can be seen that the transducer is linear to within 3-1/2 percent from 0 to 200 mmHg and exhibits a sensitivity of 27.5 microvolts per mmHg.

Temperature sensitivity was tested over a 12°C range from 22°C to 34°C. Transferring the transducer from one water bath to another over this range resulted in an output response that rose to 200 microvolts in about two seconds and then gradually returned to within 80 microvolts of the original level during a 2-minute interval while temperature of the transducer components came to equilibrium with the new environment. This represents an equivalent pressure response of 0.62 mmHg/°C for rapid temperature changes, or 0.25 mmHg/°C for changes of a very slow nature. This represents a decrease in temperature sensitivity to about 1 percent of that of the original, Model 4, transducer. The reduction in temperature sensitivity described above was obtained by using a four-active-arm strain gage bridge of matched gages. The thermistor was not needed to compensate the transducer.

The transducer was also tested for calibration sensitivity to temperature variations. The response to a 100-mmHg pressure input from the artificial artery was measured at temperatures ranging from 22°C to 36°C. The resulting shift in calibration was determined to be 1.7 percent per degree centigrade, and was linear over this range of temperature.

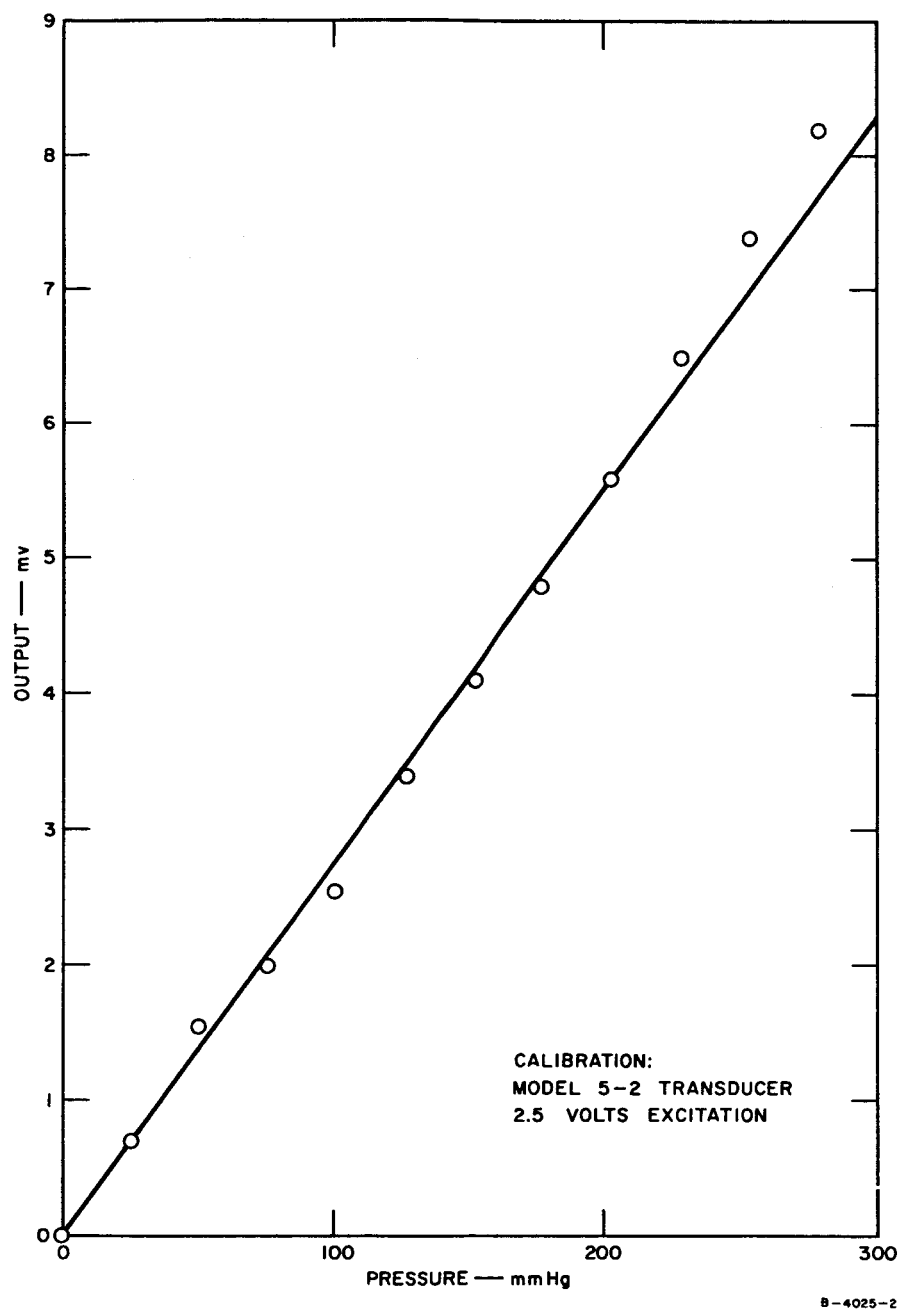


FIG. 2 TRANSDUCER STATIC CALIBRATION

B. DYNAMIC CALIBRATION

The rubber membrane apparatus was equipped with a Statham Model PM-60TC \pm 5 pressure transducer to measure dynamic pressures acting on the membrane to load the SRI blood-pressure transducer. The pressure was increased from 0 to 200 mmHg in 1/5 second, with a maximum rate of pressure rise of about 2000 mmHg/sec. Both transducer outputs were fed to an X-Y oscilloscope to obtain the dynamic calibration curve shown in Fig. 3. The reference transducer output is recorded on the X-axis, the SRI transducer output on the Y-axis.

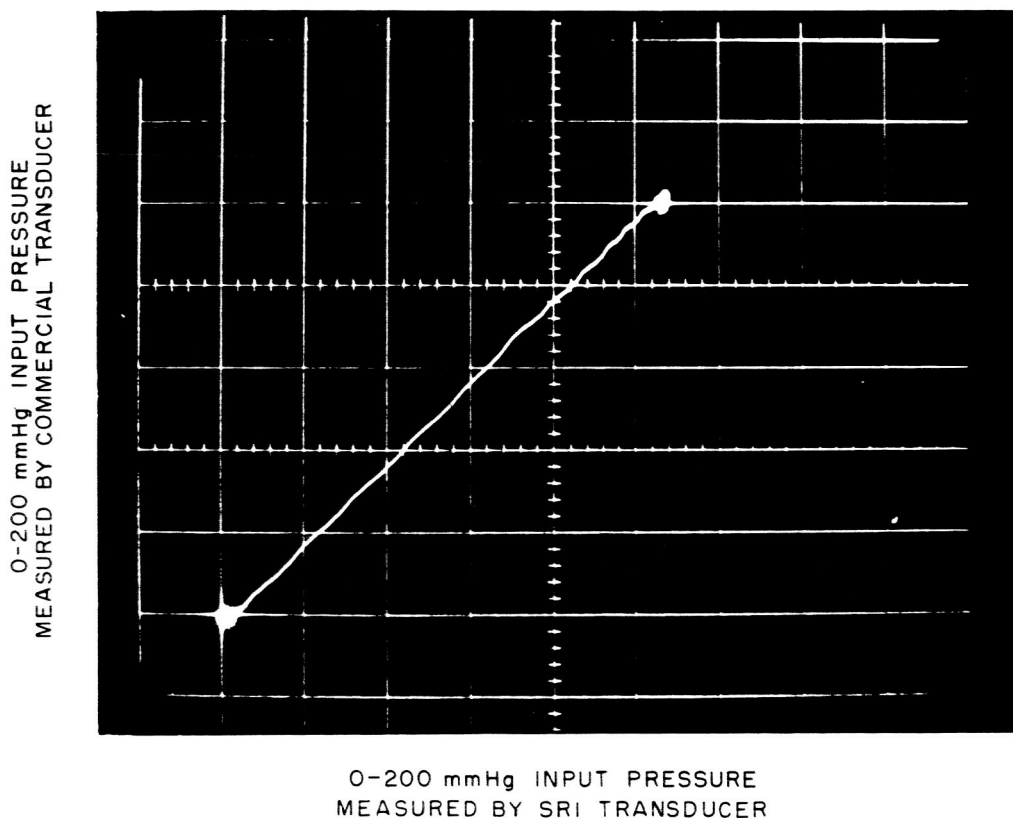


FIG. 3 DYNAMIC RESPONSE CURVE

The oscilloscope photograph presents a graphic display of the transducer's linearity under dynamic pressure loadings. Further tests of this sort have shown the existence of a creep hysteresis. Holding the pressure at some constant level such as 200 mmHg will result in a

slow drift of transducer output in which the output tends to decrease over a one-minute period to cause an eventual change of about 15 mmHg. The method of static calibration used on this transducer compensates for this effect so that the eventual blood pressure reading is correct after about the first minute of measurement. The existence of this phenomenon, however, suggests the existence of creep in the epoxy bonding of the beam to the transducer base, or of the bonding of the gages on the beam. Although not of great significance to the operation of this transducer, this phenomenon may pose a problem if the present design is extended to smaller scale with lower strain levels.

C. DRIFT TEST

The transducer was placed on the radial artery and pressures were recorded for a period of 20 minutes. At the beginning of the run, the transducer read a pressure of 115/62, while the sphygmomanometer read a pressure of 107/65. At the end of the run the transducer read a pressure of 105/52, while the sphygmomanometer read 110/67. Calibration at the end of the run revealed a zero shift of -10 mmHg and a shift in calibration of -5% (7 mm out of 150 mm test pressure). Variations in blood pressure readings during the run were attributed to shifting of the transducer over the artery. The position was re-established in order to obtain the final readings.

D. COMPARISON TO SPHYGMOMANOMETER

A series of tests was performed to determine how well the blood-pressure transducer readings compared to systolic and diastolic readings taken with a sphygmomanometer. The transducer was carefully positioned on the left radial artery to obtain the maximum pulse-wave-amplitude output. A standard sphygmomanometer and an electronic stethoscope were used to record systolic and diastolic pressure in the right arm. Both readings were taken simultaneously and recorded on a two-channel recorder. A typical recording is shown in Fig. 4. The upper trace shows cuff pressure with Korotov sounds superimposed. The lower trace is the pre-calibrated blood-pressure transducer output. The results of these tests are tabulated in Table I.

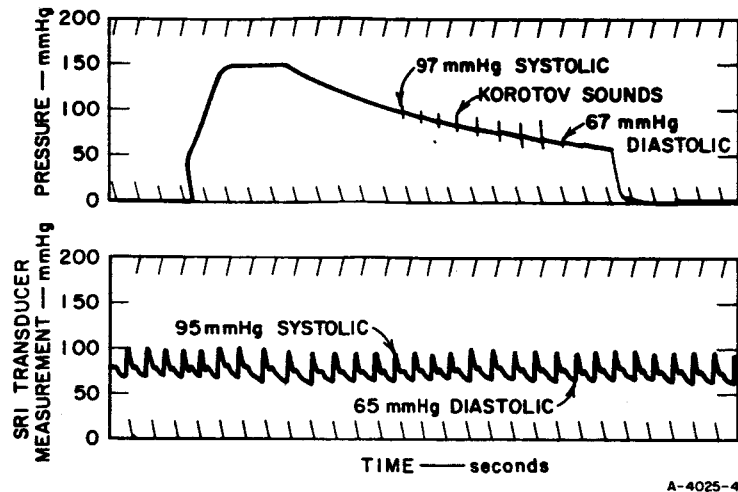


FIG. 4 TYPICAL COMPARISON TO SPHYGMOMANOMETER

The differences between the two measurement methods are attributed to the following factors: (1) error in sphygmomanometer readings, and in interpretation of stethoscope recordings, especially in the diastolic measurement; (2) difference in blood pressure between the two arms; (3) error in positioning of the transducer; and (4) size mismatch between artery and transducer rider.

During these tests, the transducer was found to be quite sensitive to position. It was necessary to hold the transducer in position rather than rely on a strap.

The technique of searching for the highest pulse output by changing position and box pressure proved to be quite practical for the original positioning of the transducer.

E. CENTRIFUGE TEST

The blood-pressure transducer was applied during centrifuge tests on October 24, 1962, at the University of Southern California. Measurement was attempted on two subjects. The initial location of the temporal artery was not difficult when the transducer was hand-held. Optimum pulse amplitudes of 60 percent of the radial pulse could be obtained on both subjects. The transducer could not be positioned as accurately

Table I

COMPARISON OF TRANSDUCER TO SPHYGMOMANOMETER

Subject	Age Years	Height	Weight lbs.	Sphygmomanometer		SRI Transducer		% Difference	
				Systolic mmHg	Diastolic mmHg	Systolic mmHg	Diastolic mmHg	Systolic	Diastolic
A	23	6'2"	170	97	67	95	65	- 2.1	- 3.0
B	30	5'9"	164	125	90	115	88	- 8.0	- 2.2
C	31	6'1"	155	125	77	125	68	0.0	- 11.7
D	27	5'8"	150	110	70	100	75	- 9.1	+ 7.1
E	35			120	87	110	85	- 8.3	- 2.3
F	28	5'9"	155	110	80	105	65	- 4.5	- 18.8
G	39	5'11"	140	115	82	112	85	- 2.6	+ 3.5
H	41	5'11"	170	112	85	107	82	- 4.5	- 3.5
I	25	6'0"	175	111	80	102	75	- 8.2	- 18.8
J	29	5'6"	125	108	80	100	83	- 1.9	+ 3.8
K	41	5'7"	140	117	67	100	67	- 14.5	0.0

under the headband as when it was hand-held, and pulse amplitudes dropped to approximately 10 percent of the radial pulse. When the helmet face mask was closed, the transducer was shifted and the pulse was lost entirely. This shifting was believed to occur when the face mask touched the headband over a large area, which led to a compressing effect as the helmet was closed and reduced the headband tension. The same loss of signal could be produced by pressing the headband against the forehead. It was not possible to obtain a signal from the first subject with the headband crossing the forehead and with the helmet closed. As a final effort, the headband was removed and replaced by a chin strap, passing over the top of the head. With this arrangement, pulses were obtained after the helmet was closed, although even here the subject's talking caused the band to shift, with a gradual loss of signal.

On the second subject the transducer was placed on the zygomatic branch of the temporal artery, and the headband applied with greater tension. The face mask was cut out to leave sufficient room for the transducer, and the subject was instructed to force his head against the helmet on the side opposite the transducer during the closing of the face piece. The face piece was then closed with no loss of signal.

The signal was recorded satisfactorily until acceleration was begun. As the acceleration increased, the pulse amplitude reduced and finally disappeared. Nothing but vibrational noise could be observed during the acceleration period (the first run was made at 2g). As the acceleration reduced, the pulse amplitude was restored and reached a maximum, diminishing again as zero g was reached. Therefore, some permanent shift occurred. The 4g run exhibited the same effects.

F. COMPARISON WITH DIRECT INTRA-ARTERIAL MEASUREMENT IN ANIMALS

Further testing of the transducer was performed by measuring the blood pressure in the arteries of a dog and a cat. An initial attempt was made to acquire the pulse of the femoral artery through the (shaved) skin. This was not possible in the cat because of the depth of the artery, its location in a non-flat region of the body, and the absence of hard underlying tissue or bone to support the artery. The dog's artery could be palpated easily and good pulses were obtained. However,

there was a severe discrepancy in pulse shape as compared to the cannula recording (see Fig. 5). The artery was exposed, and the measurement made again; the same pulse shape discrepancy existed.

Indirect pressure readings were obtained from the exposed femoral artery; direct measurements were made with a Statham pressure transducer connected to a cannula in the contralateral femoral artery. The femoral artery of the cat appeared to be approximately 0.050 inch in external diameter;

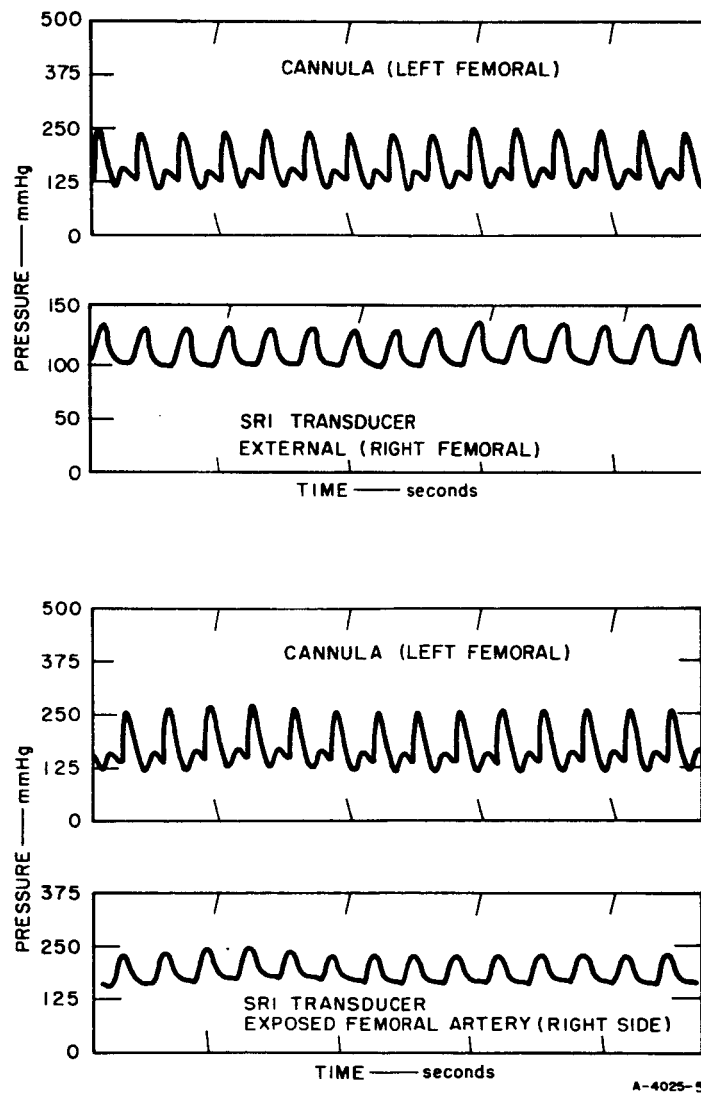


FIG. 5 BLOOD-PRESSURE RECORDINGS FROM THE FEMORAL ARTERIES OF A DOG

positioning of the transducer was therefore somewhat difficult. However, satisfactory pulses could be obtained for brief periods. The results are given below.

Pressure in Femoral Artery of a Cat

	(mmHg)	
	<u>Cannula</u>	<u>SRI</u>
Systolic	120	98
Diastolic	105	77
Mean	110	84

Readings were also obtained from the carotid artery and abdominal aorta of the cat, both of which are larger than the femoral artery. The results are summarized as follows:

<u>Artery</u>	<u>Cannula</u>	<u>SRI Transducer</u>	<u>Est. Distance from Cannulation Point</u>
Carotid	120/99 - 106	125/95 - 105	12 inches
Abdominal Aorta	120/95 - 103	125/90 - 102	4 inches

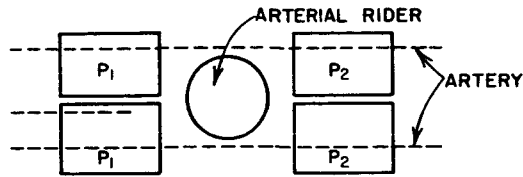
In performing the above tests on dogs, the high systolic readings and pronounced dichrotic notch displayed by the cannulation system indicated insufficiently damped response. When tested, the system displayed an overshoot of 60 percent in response to a step-pressure input of 100 mmHg and 30 millisecond rise time. The resonant frequency is 7 cps. (The pulse rate of the dog was 4 cps.) There was a decay time of 34 milliseconds. Because the cannula system did not faithfully follow the blood pressure pulses, only the mean value has significance. The complex pulse shape made the determination of mean pressure difficult. However, when variations of blood pressure were caused by stimulation of the vagus nerve and injection of epinephrine and acetylcholine, the transducer followed the variations faithfully, even though these techniques cause significant changes in vascular tone.

For future tests, improved dynamic response of the cannula system can be obtained by reducing the length of tubing between the transducer and the artery.

V THE SYSTEM CONCEPT

The positioning of the transducer has been shown to be critical to calibrated pressure readings. It is, therefore, necessary that the design of a blood-pressure measuring system include a means of positioning the transducer accurately. When the transducer is positioned manually, observation of the pulse amplitude is used as a guide; the maximum reading occurs when the transducer is in the correct position. If the transducer is to be used continuously for long periods, a means of mounting must be provided. It has been found impossible to position the transducer manually before the mounting device is applied, since the installation procedure invariably moves the transducer. Manual positioning after mounting is very difficult because of the force with which the transducer is held against the body. A satisfactory transducer mounting device should include a means of lifting off and moving the transducer without disturbing the mounting system. (The lifting off is desirable to allow the zero level of the transducer to be checked.) With such a mounting device, the transducer could be positioned after it had been mounted on the body, using pulse amplitude as a criterion. If a helmet were to be placed over the mounting system, a remote positioning control would be necessary, since placing the helmet would probably shift the transducer. The remote control of transducer position would also allow repositioning during a test or experiment if acceleration or subject movement displaced the transducer.

If position sensors could be added to the system, it would not only be possible to tell when the transducer was off the artery, but the direction of the required correction. These position sensors should detect the location of the artery with respect to the transducer rider. Small pulse sensors located immediately ahead and behind the arterial rider would provide the required information. A possible arrangement is shown in Fig. 6. The squares marked P are highly sensitive, but not necessarily linear or stable pressure sensors. If the sensors marked P_1 are connected so that their outputs oppose, and the sensors P_2 are connected the same way, there would be no output from either sensor



A-4025-6

FIG. 6 PULSE SENSOR ARRANGEMENT

when the rider was properly centered over the artery. An offset in one direction would produce a positive output from the pulse sensors, and a negative output would result from the opposite shift in position.

If the pulse sensors produce a satisfactory signal-to-noise ratio, they can be combined with the remote positioning device to form a feedback-controlled positioning system. Such a system would maintain the correct transducer position under most conditions of uniform acceleration. A shock severe enough to move the transducer rapidly, and with enough amplitude to move the sensors permanently off the artery, would defeat the system. Manual control would then be used to relocate the artery.

Since the pulse sensors are a secondary part of the over-all blood-pressure measurement system, they must be simple in design and easy to incorporate into a small transducer. At this point the use of pressure-sensitive paints appears to be the most convenient. Some limited experiments have been performed with sensors made of this material, but arterial pulses could not be obtained; this is attributed to lack of sensitivity in the paint. More sensitive paint is available, and thinner layers are required in order to achieve the required sensitivity.

VI ARTERY SURVEY

Ten of the most superficial of the large arteries were surveyed to determine their suitability for use as pressure measuring sites. There was no attempt to modify the transducer in any way to fit a particular location. Transducer box pressure was varied in the usual manner to produce the highest available pulse amplitude. Box pressure, therefore, gives an indication of the amount of tissue overlying the artery. Results of this survey are given in Table II.

Although the radial artery shows good calibration, previous experience¹ indicates that it is extremely difficult to maintain the transducer position on a mobile subject.

The femoral artery measurement shows good agreement with the sphygmomanometer, but it is not available for use on a subject who is sitting.

The brachial artery is a somewhat better location than the radial and shows fair agreement with the sphygmomanometer. However, the high box pressure required to make the measurement indicates the presence of a great deal of intervening tissue material above the artery. Since this artery is larger than the radial artery, the loss of calibration must be due to the tissue material, and would probably not be improved by modifications in the transducer.

Since readings on the two branches of the temporal artery were made with relatively moderate air box pressure, it is believed that the loss of calibration at these sites is due primarily to a size mismatch between the artery lumen and the transducer rider. This indicates that calibration could be improved by reducing transducer size. In addition, the temporal artery site has been shown to be relatively free of motion artifact disturbances, and pressure measurement at this site should not encumber the subject.

¹Final Report, NAS 2-515. X

Table II

RESULTS OF ARTERY SURVEY

Artery	Load Pressure inches H ₂ O	Sphygmomanometer		Transducer		% Error (systolic)
		Systolic mmHg	Diastolic mmHg	Systolic mmHg	Diastolic mmHg	
1. Left Radial	30	118	75	112	70	- 5.6
2. Right Femoral (supine subject)	50	120	75	110	65	- 8.3
3. Left Brachial Artery	90	110	70	95	75	- 13.6
4. Abdominal Aorta	150	120	75	75	53	- 29.2
5. Right Superficial Temporal	50	110	73	62	52	- 43.6
6. Right Facial	---	110	68	45	35	- 59.1
7. Right Occipital	---	110	65	45	30	- 59.1
8. Frontal Branch of Superficial Temporal (right side)	50	105	70	25	10	- 76.1
9. Right External Carotid	150	120	70	15	5	- 87.4
10. Common Carotid	---	Not distinguishable				

VII CONCLUSIONS AND RECOMMENDATIONS

The results of the testing performed in this project show that the newly designed transducer is superior to the first prototype as a pre-calibrated blood-pressure indicator. Because of the improved characteristics of the new transducer, the concepts which led to the design changes are considered to be valid, and indicate the direction for future improvements. These concepts are:

- (1) The rider should be smaller than the artery lumen.
- (2) The restrained artery deflection should be at least one-tenth of the deflection of the skin above the artery when limited by the area of rider aperture.

It is, therefore, recommended that a transducer be designed for use specifically on the superficial temporal artery.

Judging from the experience obtained with the present transducer, the temporal artery is from 50 to 70 percent the diameter of the radial artery. Thus the rider dimension for a temporal artery transducer must be less than 0.020 inch. A circular rider is desirable to eliminate the need for transducer alignment with the artery direction. The estimate for free skin deflection, which establishes the required stiffness of the measuring system, must be revised; the "free" deflection to be considered is the actual displacement as measured through the rider hole (between the side plates) without the rider present. Since this deflection is probably much less than the 120 microinches assumed for the free temporal artery, the allowable deflection of the rider must be correspondingly reduced to approximately one microinch.

In addition, the transducer's over-all size must be reduced, an air box must be designed for location near the ear, and a mounting system must be developed that allows the establishment and maintenance of transducer position.

The design and construction of a blood-pressure transducer system that meets these recommendations is admittedly difficult. However,

this problem is not considered to be beyond the state of the art in measurement devices, and a thorough investigation of approaches to the design of such a system is recommended.

APPENDIX

TRANSDUCER CONSTRUCTION

Figure A-1 shows the major components of the transducer in an assembly diagram.

Part No. 1 (see Fig. A-2), the strain gage beam, is made of 24 ST aluminum. It has an effective length of 0.220 inch with a thickness of 0.007 inch and a width of 0.125 inch. The beam includes the arterial rider as an integral part. Epoxy adhesive is used to mount the beam on the base plate and between the side plates. Four strain gages (MS-301-120) and one thermistor (VECO-31 A-7) are fastened to the top of the beam with epoxy cement as shown in Figs. A-1 and A-2.

Part No. 2, the base plate, is milled from 24 ST aluminum. This item forms the support structure for the entire transducer, and is shown in Fig. A-2.

Part No. 3, side plates, is also made from 24 ST aluminum. The function of these is to provide support for the beam and to hold the feed-through electrical connector groups. As shown in Fig. A-2, the feed-through connectors are formed of No. 24 AWG wire molded into position through the side plates with RTV-60 silicone rubber.

When assembly of the first three items was completed, the transducer face was polished flat to present a smooth flush surface to the skin.

After wiring the strain gages to the feed-through connectors with No. 41 AWG wire, the transducer top, Item 4, was installed using RTV-11 silicone rubber as a cement.

The air box consists of aluminum ring 1 inch OD, 1/32 wall, 7/16 high, with a lucite cover.

The diaphragm is molded of RTV-11 silicone rubber to a thickness of 0.015 inch. During the molding operation, the diaphragm is bonded to both the transducer and air box.

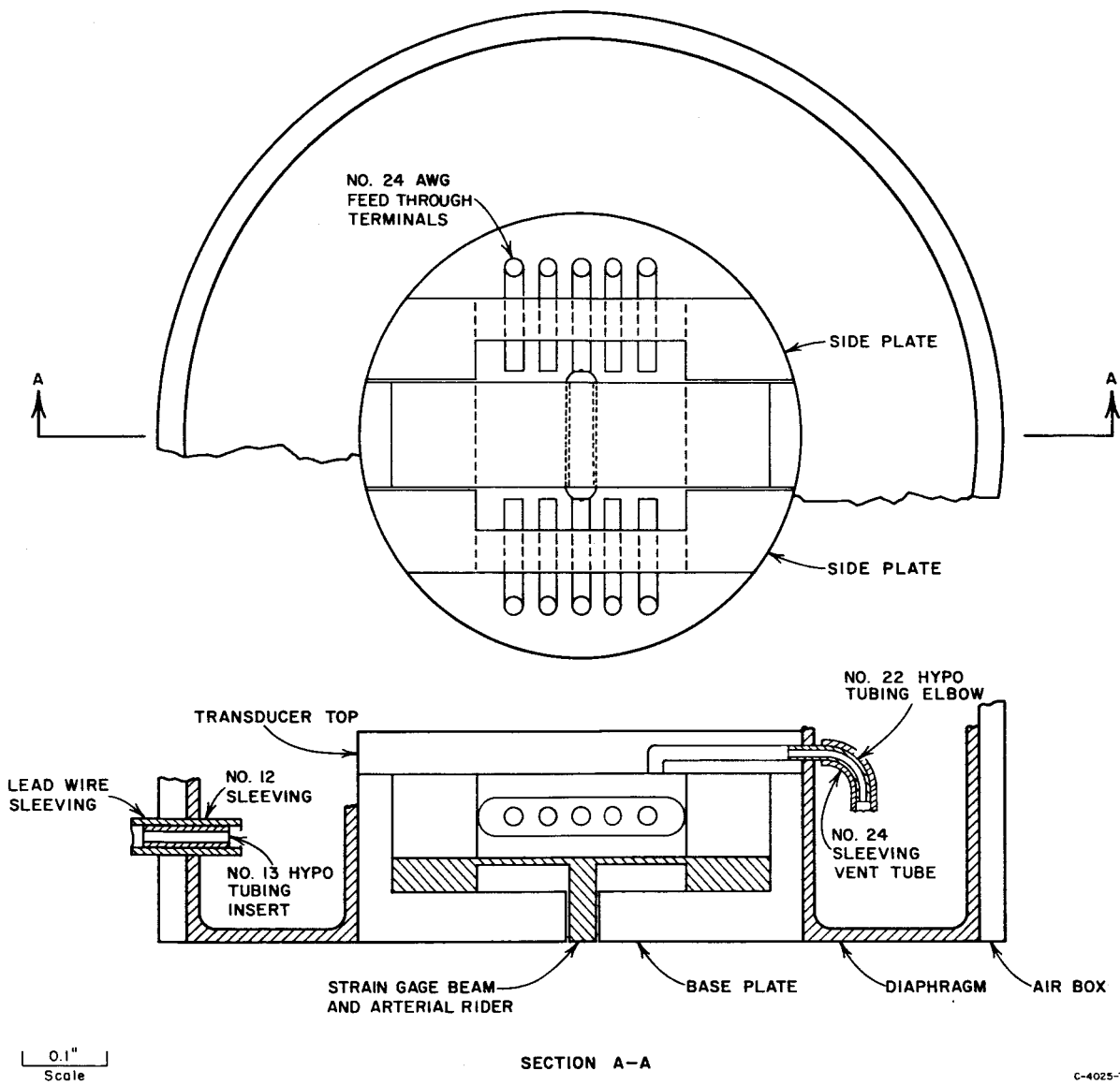


FIG. A-1 TRANSDUCER CONSTRUCTION DRAWING

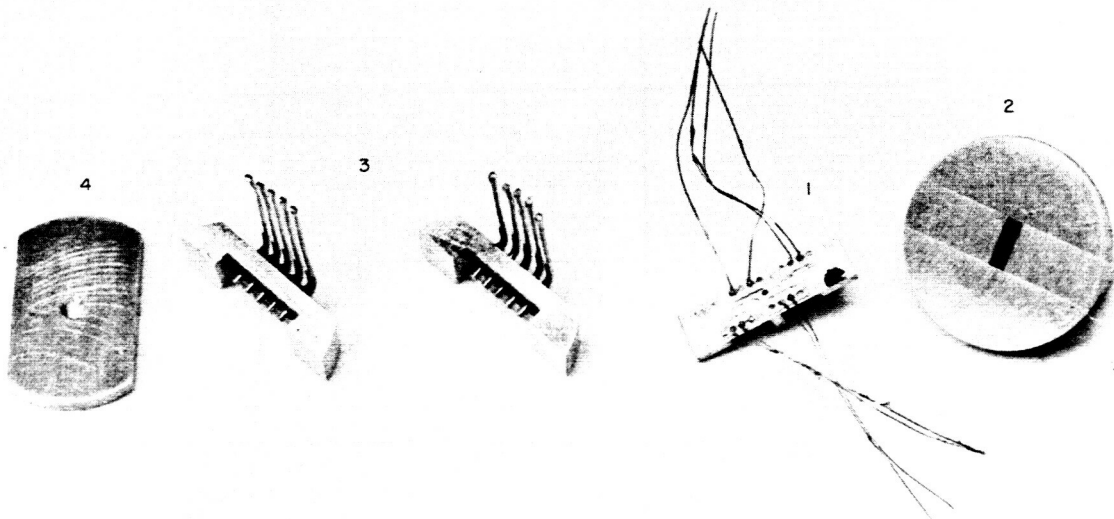
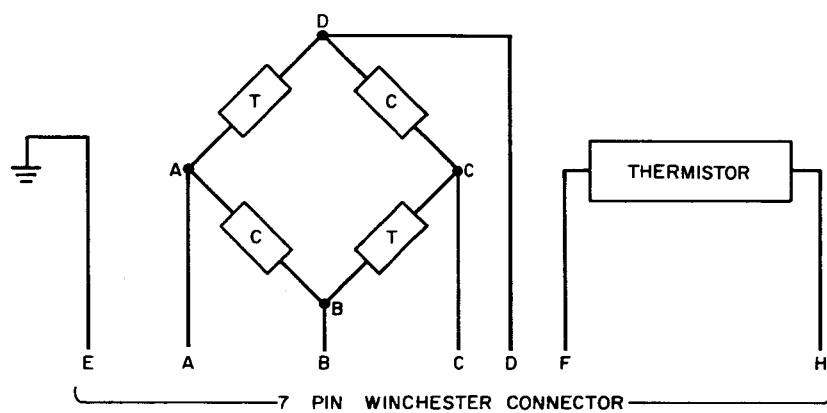


FIG. A-2 TRANSDUCER COMPONENTS - (GRID LINES 1/8 INCH APART)

A vent tube connects the interior of the transducer to atmosphere. This provides local ambient pressure as the reference for blood pressure.

Six lead wires are introduced through the air box wall. These wires are enclosed in a tube that serves as an air supply tube. Figure A-3 shows a wiring diagram of the transducer.



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FIG. A-3 TRANSDUCER WIRING DIAGRAM